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Simulating the effects of fatigue and pathologies in gait

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ABSTRACT

In this study, we propose a new method to simulate the effect of fatigue and pathologies in human gait motion. The method is based on Angular Momentum inducing inverted Pendulum Mode (AMPM), which is the enhanced version of 3D linear inverted pendulum mode that is used in robotics to generate biped locomotion. By importing the gait motion captured using a motion-capture device, the value of AMPM-parameters that define the trajectory of the center of mass and the angular momentum are calculated. By minimizing an objective function that takes into account the fatigue and disabilities of muscles, the original motion is converted to a new motion. Since the number of parameters to define the motion is small in our method, the optimization process converges much more quickly than in previous methods.

1. INTRODUCTION

Computer simulations based on musculoskeletal models are often used to analyze the role of specific muscles during motions. Many researchers have analyzed the contribution of some specific muscles to the performance of certain motions by using optimization methods based on forward dynamics. For example, Pandy and Zajac (1991) analyzed the role of biarticular muscles in maximal jumping, and they concluded that biarticular muscles contribute to jumping performance by redistributing segmental energy within the musculoskeletal system but do not contribute to the energy of jumping. Neptune et al. (2001) analyzed the role of the plantarflexor muscles during gait, and they calculated the degree to which these muscles contribute to propelling the trunk in the forward direction. Human gait motions have also been simulated by combining forward dynamics and optimization (Yamaguchi and Zajac, 1990; Anderson and Pandy, 2001; Neptune et al., 2001). Although these researches have succeeded in simulating realistic normal gait, there are several shortcomings with these methods that make it difficult to apply them for the diagnosis of patients due to the following reasons:

1. First of all, researchers must provide a good initial guess of the activation data to prevent the optimization process from getting caught into some local minima. In most of the cases, this is done through trial-and-error by researchers. This is quite time consuming, and the researcher needs special experience to determine the good set of muscle activations to generate realistic gait motion. In addition, since this process is done completely manually, the motion can become subjective to the researcher.
2. Next, since all the time sequence data for the muscles must be determined through the optimization, costly computation is required until the result is converged to the optimal value. Neptune (1999) reported that more than thousands of iterations are needed before the optimized value is obtained.
3. Third, although balance-keeping is one of the most important factors for gait motion, there was no explicit way to describe the balanced motion in these researches. As a result, the search space for the parameter is much larger since it also includes those parameter values that lead to unbalanced motion.

In robotics, biped locomotion of humanoid robots is one of the most exciting topics these days. Many researchers have developed humanoid robots that are capable of performing biped gait motion (Kajita et al., 1992, 2001). A method commonly used to generate gait motion is the 3D Linear Inverted Pendulum Mode (3DLIPM) (Kajita et al., 1992, 2001). The great advantage of the 3DLIPM approach is that the trajectory of the center of mass (COM) can be written in an explicit form. The 3DLIPM approach is a hierarchical approach, in which the abstract motion such as the trajectory of COM is determined first, and the details of the motion such as the kinematic data are calculated later. The advantage of the hierarchical approach is that the controller does not have to waste time determining low level control signals such as the torque exerted at the joints or the muscle activation data. With the hierarchical approach, it is possible to simplify complex models of humanoid robots that have too many degrees of freedom (DOF) to be controlled directly.

In addition, as the motion of the COM is explicitly controlled, the balance of the humanoid robot is assured in the feedforward stage. However, no angular momentum has been generated by the 3DLIPM since it assumes that the COM is a mass point and that the ground force vector always passes through the COM of the system (Figure 1(a)).

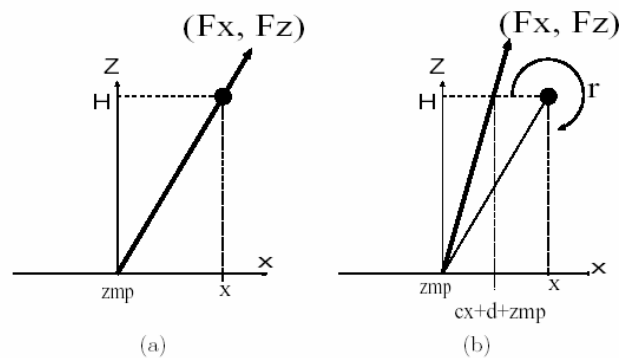


Figure 1. The standard 3DLIPM (a) and AMPM (b). The 3DLIPM restricted the ground force to pass through COM while the AMPM allows the horizontal component of the ground force to have linear relationship with the position of the COM.

It is known that angular momentum is consistently generated around the COM when humans walk. The amount of angular momentum gets even larger as the motion involves larger bending of the thorax, which can be often observed in pathological gait. It is therefore difficult to model human gait motion by using 3DLIPM. To represent trajectory of COM and angular momentum of human gait motion in an explicit form, we have extended the 3DLIPM model so that angular momentum can be induced by the ground reaction force (Kudoh and Komura, 2003). This model is called *Angular Momentum inducing inverted Pendulum Model* (AMPM). In this study, we propose a new hierarchical approach to calculate human gait motion using AMPM. By combining AMPM with the musculoskeletal model, it is even possible to calculate pathological gait. The effects of deactivating the gluteus medialis muscles were simulated and compared with the motion of patients with similar disabilities.

The trajectory of the COM and angular momentum is first calculated by using AMPM. Next, the trajectories of the generalized coordinates including the position of the pelvis and the joint angles of the whole body are determined using inverse kinematics. After determining the muscles of the body to be deactivated, an objective function based on the time series data of the force exerted by this muscle is formed. By searching for an optimal set of AMPM parameters that minimizes the objective function defining the gait motion, the pathological gait motion is calculated.

The method proposed in this study has the following advantages compared with previous methods:

- Since the gait motion is described by the AMPM parameters, there is no need to specify all the input parameters of the muscles. As a result, the computational cost for the optimization is much less than previous methods.
- As the balance of the human body model is explicitly kept by using the AMPM model, the optimizer only needs to search for the optimal set of parameters in terms of muscle force, and hence does not need to go through a large number of trials to generate the balanced motion.

2. METHODS

In this section, the method to simulate the effects of pathologies and fatigue during human gait is explained. First we review the Angular Momentum inducing inverted Pendulum Model (AMPM) (Kudoh and Komura, 2003) and also explain further enhancements so that it can handle imported human motion data. Next, the details of the musculoskeletal model we used in this research are explained. Finally, the method to simulate the effects of pathologies and fatigue by optimization is explained.

2.1 Angular Momentum inducing inverted Pendulum Model

The AMPM enhances the 3DLIPM in a way the ground force vector is calculated to be not only parallel to the vector connecting the ZMP and the COM; its horizontal element can be linearly correlated to the ZMP-COM vector (Figure 1(b)). As a result, rotational moment will be generated by the ground force. In the previous studies, we assumed the ground force vector to be linearly correlated to the position of the COM:

$$\ddot{x} = Ax + B \quad (1)$$

As we assume the single support phase of human gait follows this rule, the parameters A and B can be calculated using the position and acceleration of the COM at the lift-off and heel-strike during gait. However, when actually we substitute data into this equation, the trajectory of the acceleration does not match well with those by humans. As shown in Figure 2, oscillatory gap is observed between the acceleration of the COM and that generated by AMPM.

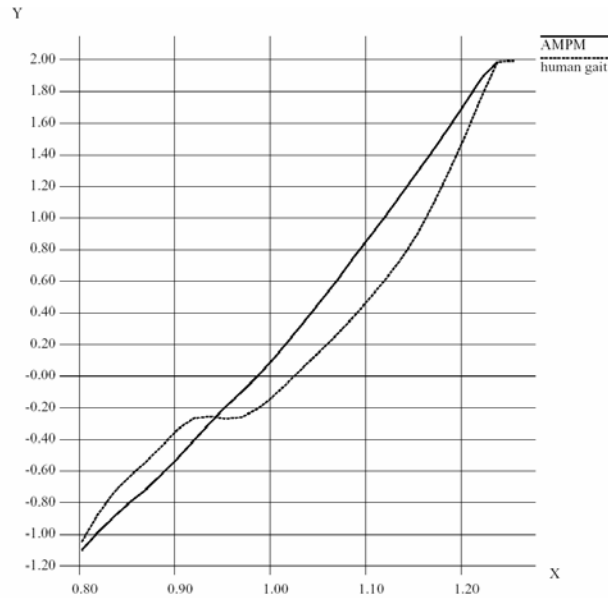


Figure 2. Acceleration of the COM during gait (dashed line) and that by AMPM (bold line).

In order to decrease the gap between the trajectory by the AMPM and the trajectory of the COM during gait, we propose to add trigonometric functions in the right side of Equation (1) as follows:

$$\ddot{x} = Ax + B + C \cos \omega t + D \sin \omega t \quad (2)$$

The parameters C , D , ω are decided in a way that trajectory of the differential equation matches well with the trajectory of the COM. Equation (2) has the explicit solution as follows:

$$x = -\frac{B}{A} + C_1 e^{-\sqrt{A}t} + C_2 e^{\sqrt{A}t} - \frac{\cos \omega t + D \sin \omega t}{A + \omega^2} \quad (3)$$

where C_1 , C_2 are constant values which can be calculated from the terminal conditions. In this study, we make the following two assumptions through observation of human motion:

- The position of the ZMP is linearly correlated to the position of the COM. Therefore, the position of the ZMP can be defined by $(zmp, 0) = (cx + d, 0)$, where (c, d) are constant parameters.

- The height of the COM can be expressed by the following equation:

$$z_g = H + z_h \sin \frac{t}{T}$$

where H is a constant value, T is a half cycle of gait motion, z_h is the amplitude of the motion, and z_g is the height of the COM.

Then, the ground force vector can be written as

$$F_x = m\ddot{x} = m \left(AC_1 e^{-\sqrt{A}t} + AC_2 e^{\sqrt{A}t} + \frac{\omega^2}{A + \omega^2} (\cos \omega t + D \sin \omega t) \right)$$

$$F_y = m \left(g - \frac{1}{T^2} \sin \frac{1}{T} \right)$$

where m is the mass of the system. The rotational moment r around the y -axis can be calculated by

$$r = ((1-c)x + d)F_y - (H + z_h \sin \frac{t}{T})F_x$$

Using this moment, it is also possible to calculate the angular momentum generated by the rotational momentum in an explicit form, although we cannot list it here due to its length. This form was calculated using the software *Mathematica*. AMPM is used to represent the motion of the COM/ZMP/angular momentum in the frontal and sagittal plane. Then, the generalized coordinates of the body are then calculated using inverse kinematics (Kudoh and Komura, 2003).

2.2 Musculoskeletal Model

To simulate the motion by pathological patients, it is necessary to prepare a physiological model of the human body. The musculoskeletal model developed by Delp (1990) was used in this study. This data includes the attachment sites of 43 muscles on each leg and physiological parameters such as the tendon slack length, optimal fiber length, pennation angles, and maximum exertable force. Muscles are attached only to the legs, and no muscles are put on the upper half of the body. The upper half of the body is composed of the thorax, head, upper arms, lower arms, and hands.

However, only the thorax (3 degrees of freedom (DOF)) and the upper arms (3 DOF each) are allowed to move among these segments. The lower half of the body is composed of the pelvis, and the femur, tibia, patella, talus, calcaneous, and toes in each leg. The joints of each leg are composed of a 3-DOF gimbal joint (hip joint) and 1-DOF joint (knee, ankle, calcaneous, and metatarsal joint). Therefore, the total degrees of freedom of the body, including the 6DOF of the pelvis in the global coordinate system, is twenty nine.

The musculotendon model of Hill (1938) is used, and parameter values were derived according to Delp (1990). The musculotendon model is composed of three elements: a contractile element (CE, representing all the muscle fibers), a parallel elastic element (PEE, representing all connective tissues around the muscles fibers), and a series elastic element (SEE, representing all series elasticity, including tendons). At each time step, the musculotendon length was determined from the posture (i.e., as a function of joint angles). Thereafter, it is possible to calculate the maxima and minima that bound the muscle force at each time step:

$$F_m^{\min} \leq F_m(t) \leq F_m^{\max} \quad (4)$$

where $F_m(t)$ is the musculotendon force of the m -th muscle, F_m^{\min} and F_m^{\max} are the minimum and maximum force developed by this muscle, which are determined by their force-length-velocity properties:

$$F_m^{\min} = f(F_m^0, l_m, v_m, 0)$$

$$F_m^{\max} = f(F_m^0, l_m, v_m, 1)$$

where function $f(F_m^0, l_m, v_m, a_m)$ is the force-length-velocity surface assumed in the musculoskeletal model (Zajac, 1989), F_m^0 is the maximum force constant calculated from the average cross-sectional area of the muscle, l_m is the length and v_m is the velocity of the m -th muscle being contracted, and last parameter (0 and

1) in these equations is the activation level of the muscle a_m that determines the amount of force exerted by the contractile element ($0 \leq a_m \leq 1$).

2.3 Calculating the effects of fatigue and pathologies

Torque $\tau_j(t)$ developed at joint j is theoretically generated as follows by the muscles crossing the joint:

$$\tau_j(t) = \sum_m F_m(t) r_{m,j}, j = 0, \dots, n_{dof} \quad (5)$$

where $r_{m,j}$ is the moment arm of muscle m about the j -th joint axis, and n_{dof} is the number of degrees of freedom whose torque are assumed here to be generated by the muscles. They include the flexion/extension, adduction/abduction, and rotation at the hip, flexion/extension at the knee, and plantarflexion/dorsiflexion at the ankle, and therefore, by taking both legs into account, $n_{dof} = 10$.

The muscle forces at each time step can be calculated by minimizing the muscle stress (Crowninshield and Brand, 1981):

$$J = \sum_m \left(\frac{F_m(t_i)}{F_m^0} \right)^2 \quad (6)$$

where n_m is the total number of muscles ($n_m = 43 \times 2 = 86$). J was minimized using quadratic programming, which is a method to minimize a quadratic form while satisfying linear equality and inequality constraints. In summary, the muscle forces can be calculated by minimizing Equation (6) while using Equations (5) and (4) as constraints.

To describe the process to calculate the pathological gait, a new function is defined here that summarize all the processes explained previously, including the AMPM, inverse kinematics, inverse dynamics and static optimization:

$$F_m = f_m(p, t) (m = 0, \dots, n_m - 1) \quad (7)$$

where p is the vector composed of parameters defining the gait motion by the AMPM, and F_m is the force by muscle m calculated by minimizing Equation (6). To simulate the effect of weakening muscle m , the following criterion is minimized until it is smaller than the threshold value:

$$J_m = \int_0^T [f_m(p, t) + \alpha(p, t)] dt \quad (8)$$

where T is one cycle of the gait motion, and $\alpha(p, t)$ is a penalty function based on the external torque that has to be applied to the body to assist the musculoskeletal model accomplish the motion when there is no solution found for the static optimization problem at time t (Komura et al, 2000; Komura and Shinagawa, 2001). The penalty function helps to avoid the motion to converge to one that is not feasible by the musculoskeletal model. This optimization is done by sequential quadratic programming.

3. EXPERIMENTAL DATA ANALYSIS

To examine the validity of the method proposed in this paper, we have generated normal and pathological human gait motion using our method, and compared the kinematical and dynamical data with those by humans.

First of all, the normal gait motion that was captured using a VICON motion capture system was imported, and the AMPM parameters were calculated. Next, by minimizing an objective function based on the gluteus medialis that has the form of Equation (8), the effects of weakening the gluteus medialis during gait were simulated. As the optimization proceeds, features known as lateral trunk bending appears in the motion. The trunk swings from one side to the other, producing a gait pattern known as waddling. During the double support phase, the trunk is generally upright, but as soon as the single support phase begins, the trunk leans over the support leg, returning to the upright attitude again at the beginning of the next double support phase.

The trajectory of the gait motion before and after the optimization is shown in Figure 3 (a) and (b) as well. The objective function converges to the optimal value within one hundred iterations.

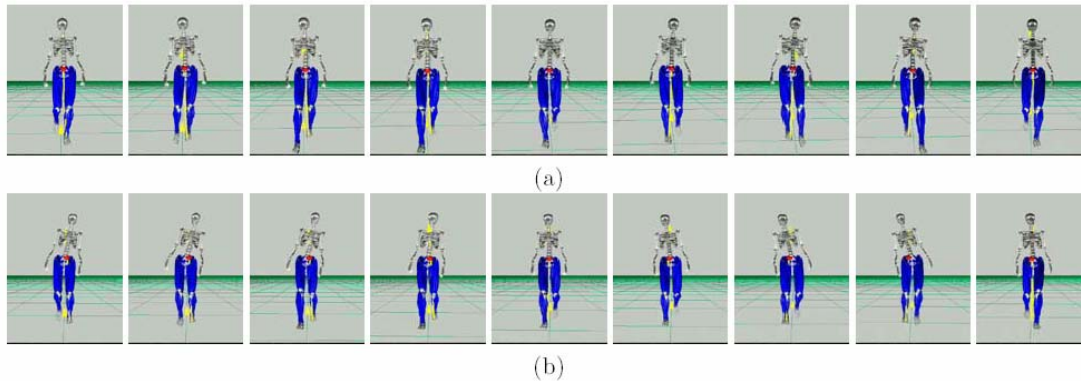


Figure 3. The trajectory of the AMPM-generated motion before (a) and after (b) optimization.

Next, in order to simulate the effect of fatigue, the maximum amount of force exerable by the muscles that are used during gait were gradually decreased. As a result, the strides of the gait gradually decreased as shown in Figure 4.

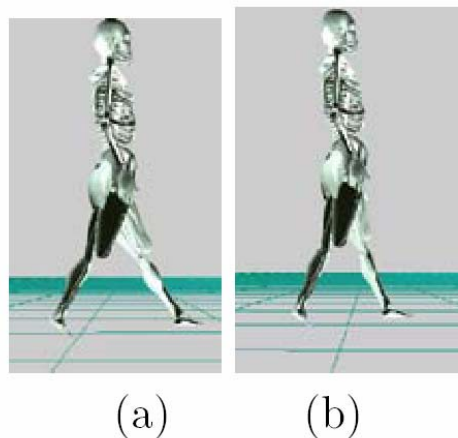


Figure 4: Simulating the effect of fatigue during gait. The strides before the optimization (a) is shortened after reducing the upper bound of the muscle force that is used during gait (b).

4. DISCUSSIONS

A method to calculate pathological gait by combining the musculoskeletal system and the inverted pendulum mode was proposed in this research. The motion is planned by the AMPM, and it is evaluated by using the musculoskeletal model. As the number of the parameters to determine the motion is less than previous methods, the optimization process converges more quickly than previous methods.

In our experiment, by minimizing the amount of force exerted by the gluteus medialis during the gait motion, the waddling gait motion was automatically induced. This is a natural phenomenon, as it is known that weak abductor muscles cause waddling. Waddling is a well-known abnormality of gait motion which is caused not only by weak abductor muscles, but also by congenital dislocation of hip joints and pain in the joints. Waddling reduces the torque and the bone-on-bone contact force at the hip of the support leg during the single support phase. The muscle force history of the musculoskeletal model performing normal gait and waddling gait calculated using static optimization are shown in Figure 5.

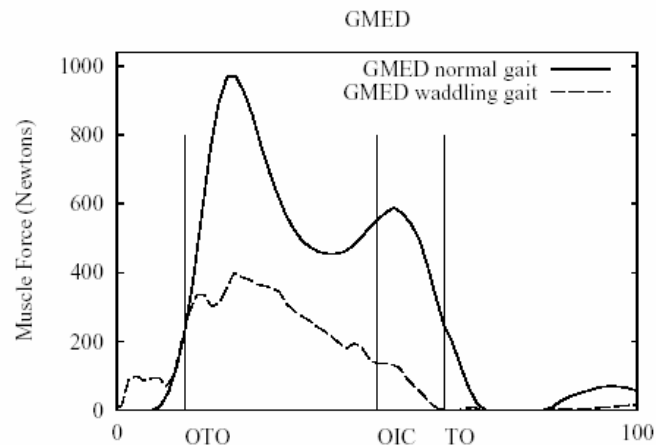


Figure 5: *The gluteus medialis (GMED) force calculated by static optimization method using human motion by a healthy subject (bold line) and by a patient with congenital dislocation (dashed line).*

Compared with dynamic optimization methods that have recently been used in biomechanics to simulate human gait and estimate muscle force, the method proposed in this study has the following advantages.

1. Our method to calculate human gait motion requires only a small number of parameters that define the motion of the AMPM. For example, if the musculoskeletal system is composed of 86 muscles, and during the gait cycle each muscle is allowed to change its input value 100 times, the total number of parameters that define the whole motion becomes 8,600. This means at least this number of iterations are needed for calculating the derivative of the criteria of optimization. Although techniques such as using the same input signals for muscles with similar roles (Pandy et al., 1990), or controlling muscles by “Bang-Bang” methods to reduce the search space (Yamaguchi and Zajac, 1990), are used, the number of parameters still remains very large. On the other hand, when using the proposed method, the number of parameters is less than ten. Since the number of the parameters is small, the optimization converges much more quickly than dynamic optimization methods. Comparing the amount of computation needed, our method requires less than one tenth of the forward dynamics approach.
2. Since the balance of the motion is assured by the algorithm of the motion generation, the optimizer does not have to spend further effort in keeping the balance through the gait cycle. This is one of the most important contributions of the paper, because when using optimization methods based on forward dynamics to simulate bipedal gait, most of the iterations are done for unbalanced motions tumbling down onto the ground. If the system can avoid evaluating such motions, the search for the optimal motion can be done much more efficiently.
3. It is not necessary to search manually the initial guess of the muscle-activation parameters that decide the motion, as the parameters that define the gait based on the AMPM are intuitive kinematic parameters such as the position and velocity of the COM. As explained in the Introduction, when using optimization methods based on forward dynamics, a good initial guess of the activation data must be provided in order to prevent the optimization process from getting caught into some local minima. This is quite time consuming, and the researcher needs special experience to determine the good set of muscle activations to generate realistic gait motion. In addition, as such process is done completely manually, the motion can become subjective to the researcher.

5. SUMMARY AND FUTURE WORK

In this study, we have proposed a new method to simulate the gait motion when muscles are deactivated. The method is based on the AMPM which is an enhanced model of the 3DLIPM that is often used in robotics to generate gait motion. Compared with previous methods, the proposed method is simple, and the computational cost is much smaller. As further work, we are planning to fit the AMPM model into human motion data, and simulate the effects of physiological disabilities and also the process of pathology for rehabilitation use.

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